REGULARIZED VARIABLE DENSITY SENSE

DESCRIPTION

The following relates to the magnetic resonance arts. It finds particular application in magnetic resonance imaging employing non-Cartesian sampling of k-space, and in magnetic resonance imaging employing variable density sensitivity encoding (SENSE), and will be described with particular reference thereto. However, it also finds application in other magnetic resonance applications such as magnetic resonance spectroscopy.

In the SENSE magnetic resonance imaging technique, a plurality of coils are used to simultaneously acquire magnetic resonance imaging data of a common imaging slice or volume. To acquire imaging data quickly using the SENSE technique, each coil undersamples k-space. The undersampled imaging data acquired by each coil are reconstructed into corresponding folded images of the slice or volume. The folded images are unfolded based on sensitivity characteristics of the coils to produce an unfolded image of the slice or volume. The SENSE imaging technique is described, for example, in Pruessmann et al., Magnetic Resonance in Medicine 42, pp. 952-962 (1999) and in Hajnal et al., U.S. patent no. 6,380,741.

In conventional SENSE imaging, k-space is uniformly sampled. In variable density SENSE, the sampling is not uniform; rather, fewer k-space samples are skipped at or near a center of k-space while more k-space samples are skipped away from the center of k-space. As with conventional SENSE, the undersampled imaging data acquired by each coil is reconstructed into corresponding folded images. The folded images are unfolded using a suitable unfolding algorithm, such as by solving the matrix equation:

$$m(k_y) = S(r_y, k_y) \, p(r_y) \qquad\qquad (1),$$

where k_y indexes the Fourier encoding steps, r_y indexes unfolded image position, $m(k_y)$ contains folded image data, $p(r_y)$ corresponds to the unfolded image, and $S(r_y, k_y)$ corresponds to a sensitivities matrix that contains the sensitivities of the coils along the column and the corresponding Fourier encoding steps. The unfolded image term p contains N true pixel values of the unfolded image. The folded images term m contains CN/R

measured values where C is the number of coils and R is a time reduction factor corresponding to the undersampling. Equation (1) is solved for $p(r_y)$ to compute the unfolded image.

To acquire imaging data quickly using the variable density SENSE technique, the undersampling is increased. However, if the k-space sampling is too sparse, Equation (1) becomes underdetermined or close to singular. This leads to computational difficulties and can produce artifacts in the unfolded image. To address this problem, the unfolding can be regularized using *a priori* knowledge. Regularization depends upon the nature and quality of the *a priori* knowledge. Typically, this additional information is provided by acquiring additional magnetic resonance signal information which is combined with the variable density SENSE data during the unfolding.

In spite of such regularization techniques, the processing of undersampled variable density SENSE imaging data to efficiently produce an unfolded image with reduced artifacts remains a challenging problem. Image quality improvement gained by including the *a priori* knowledge may be offset by longer imaging sessions needed to acquire the *a priori* knowledge. Moreover, subject motion or other changes occurring between acquisition of the *a priori* knowledge content and the variable density SENSE data may cause a spatial misregistration or other mismatch between the *a priori* knowledge and the folded images.

The present invention contemplates an improved apparatus and method that overcomes the aforementioned limitations and others.

According to one aspect, a magnetic resonance imaging method is provided. Variable density sensitivity encoded data is acquired with a higher density at and adjacent a center of k-space and with a lower, undersampled density away from the center of k-space. One or more regularization images are constructed from the higher density portion of the variable density sensitivity encoded data disposed at and adjacent the center of k-space. The variable density sensitivity encoded data are reconstructed into an unfolded reconstructed image. The reconstructing includes: reconstructing the higher and lower density variable density sensitivity encoded data into a plurality of folded images; and unfolding the folded images to form the unfolded image using the one or more regularization images.

According to another aspect, a magnetic resonance imaging apparatus is disclosed. A plurality of radio frequency coils acquire variable density sensitivity encoded data that is undersampled at least away from the center of k-space. A reconstruction processor for each coil reconstructs: a regularization image reconstructed from a higher density portion of the variable density sensitivity encoded data disposed at or near a center of k-space acquired by that coil; and a folded image reconstructed from the variable density sensitivity encoded data acquired by that coil. An unfolding processor unfolds the folded images. The unfolding is regularized by the regularization images.

One advantage resides in reconstructing undersampled variable density SENSE imaging data with improved speed.

Another advantage resides in reconstructing undersampled variable density SENSE imaging data with minimal artifacts.

Yet another advantage resides in an ability to process undersampled variable density SENSE imaging data acquired with a variety of k-space sampling trajectories such as radial or spiral sampling.

Numerous additional advantages and benefits will become apparent to those of ordinary skill in the art upon reading the following detailed description of the preferred embodiments.

The invention may take form in various components and arrangements of components, and in various process operations and arrangements of process operations. The drawings are only for the purpose of illustrating preferred embodiments and are not to be construed as limiting the invention.

FIGURE 1 diagrammatically shows a magnetic resonance imaging system implementing the variable density SENSE magnetic resonance imaging described herein. In FIGURE 1, the magnetic resonance imaging scanner is illustrated with about one-half of the scanner cut away to reveal internal components of the scanner and to reveal an associated imaging subject and SENSE coil array disposed in the scanner bore.

FIGURE 2 plots an example k-space sampling density profile for variable density SENSE imaging.

FIGURE 3 diagrammatically shows radial k-space sampling trajectories for variable density SENSE imaging.

FIGURE 4 diagrammatically shows a spiral k-space trajectory for variable density SENSE imaging.

With reference to FIGURE 1, a magnetic resonance imaging scanner **10** includes a housing **12** defining a generally cylindrical scanner bore **14** inside of which an associated imaging subject **16** is disposed. Main magnetic field coils **20** are disposed inside the housing **12**. The main magnetic field coils **20** are arranged in a generally solenoidal configuration to produce a main B_0 magnetic field directed along a central axis **22** of the scanner bore **14**. The main magnetic field coils **20** are typically superconducting coils disposed inside in cryoshrouding **24**, although resistive main magnets can also be used. Moreover, the scanner **10** may include additional access openings other than the ends of the cylindrical scanner bore **14** for accessing the imaging subject **16**. For example, rather than a closed solenoidal configuration having a closed generally "O"-shaped cross-section, a more open generally "U"-shaped cross-sectional magnet can be employed.

The housing **12** also houses or supports magnetic field gradient coils **30** for selectively producing magnetic field gradients parallel to the central axis **22** of the bore **14**, along in-plane directions transverse to the central axis **22**, or along other selected directions. Optionally, the housing **12** further houses or supports a birdcage radio frequency body coil **32** for selectively exciting and/or detecting magnetic resonances. A sensitivity encoding (SENSE) coil array **34** disposed inside the bore **14** includes a plurality of SENSE coils, specifically four SENSE coils in the example coil array **34**, although other numbers of coils can be used. The coils of the coil array **34** have different sensitivities to the magnetic resonance signal, thus enabling SENSE encoding. In one embodiment, the coil array **34** is an array of surface coils disposed close to the imaging subject **16**. Rather than a coil array, other radio frequency antennae that have a suitable spatial sensitivity profile can be used for acquiring the SENSE data, such as antennae defined by porting the received radio frequency signal from a plurality of spatially separated ports of a radio frequency coil.

The SENSE coil array **34** can be used for receiving magnetic resonances that are excited by the optional birdcage or other whole body coil **32**, or the magnetic resonances can be both excited and received by the SENSE coil array **34**. The housing **12** typically includes a cosmetic inner liner **36** defining the scanner bore **14**.

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The main magnetic field coils **20** produce a main magnetic field B$_o$. A magnetic resonance imaging controller **40** operates magnet controllers **42** to selectively energize the magnetic field gradient coils **30**, and operates a radio frequency transmitter **44** coupled to the radio frequency coil **32** as shown, or coupled to the SENSE coil array **34**, to selectively energize the radio frequency coil or coil array **32, 34**. By selectively operating the magnetic field gradient coils **30** and the radio frequency coil or coil array **32, 34** magnetic resonance is generated and spatially encoded in at least a portion of a selected region of interest of the imaging subject **16**. By applying selected magnetic field gradients via the gradient coils **30**, a selected k-space trajectory is traversed, such as a Cartesian trajectory, a plurality of radial trajectories, or a spiral trajectory. During traversal of the selected k-space trajectory, the magnetic resonance imaging controller **40** operates a radio frequency receiver **46** coupled to the SENSE coil array **34** to read out k-space samples using the coils of the SENSE coil array **34**. The k-space samples are stored in a k-space memory **50**.

The stored k-space samples are sorted by the receive coil of the array which received each sample. The k-space sampling is variable, with higher density k-space sampling at or near a center of k-space, and an undersampled lower density k-space sampling away from the center of k-space. A reconstruction processor **52** reconstructs the undersampled k-space data spanning all of k-space acquired by each coil into corresponding folded images that are stored in a folded images memory or buffer **60**. Additionally, the portion of k-space at or near k-space center acquired by each coil at a higher sampling density is reconstructed by the reconstruction processor **52** into corresponding low resolution regularization images that are stored in a regularization images memory or buffer **62**.

An unfolding processor **66** unfolds the folded images **60** using an unfolding algorithm that is regularized by the regularization images **62**. In one suitable approach, the folded images **60** are unfolded by minimizing an error parameter δ:

$$\delta^2 = \left\| m(k_y) - S(r_y, k_y)\, p(r_y) \right\|^2 + q^2 \left\| I_{reg}(r_y) - p(r_y) \right\|^2 \qquad (2),$$

where k$_y$ indexes the Fourier encoding steps, r$_y$ corresponds to the image position at index y, m(k$_y$) corresponds to the folded images, p(r$_y$) corresponds to the unfolded image,

$S(r_y,k_y)$ corresponds to a sensitivities matrix that contains the sensitivities of the coils along the column and the corresponding Fourier encoding steps, and $I_{reg}(r_y)$ corresponds to a regularization image constructed from low resolution images stored in the regularization images memory **62**. The unfolded image term p contains N true pixel values of the unfolded image. The folded images term m contains CN/R measured values where C is the number of coils and R is a time reduction factor corresponding to the undersampling.

In one embodiment, the regularization image $I_{reg}(r_y)$ is constructed by combining the low resolution regularization images acquired by the various receive coils. The low resolution regularization images can be combined using a sum-of-squares process to produce a composite image. In another embodiment, a single low resolution regularization image acquired by one of the receive coils is used as the regularization image. In this embodiment, the selected image was preferably acquired by a receive coil that has a largely uniform sensitivity over the field of view.

The operator $\|x\|$ in Equation (2) denotes the Euclidean norm of x. Equation (2) includes unfolding term $\|m(k_y)-S(r_y,k_y)p(r_y)\|^2$ that is indicative of fidelity of the unfolded image to the folded images, and regularization term $\|I_{reg}(r_y)-p(r_y)\|^2$ that is indicative of fidelity of the unfolded image to the one or more regularization images **62**. The parameter q is a regularization weighting parameter determining the relative influences of the unfolding term and the regularization term. It will be recognized that if q=0 then unregularized Equation (1) is recovered.

In one approach for evaluating Equation (2) to obtain the unfolded image, the unfolding processor **66** minimizes Equation (2) with respect to the unfolded image term $p(r_y)$. This minimization is preferably done analytically according to:

$$p = I_{reg} + S^H \left(S S^H + q^2 \right)^{-1} \left(m - S I_{reg} \right) \tag{3},$$

where the superscript "H" denotes the transposed complex conjugate of S, the functional dependencies on k_y and r_y indicated in Equation (2) have been omitted. Equation (3) can be further optimized with respect to the regularization weighting parameter q, for example by performing a least squares minimization of the image term p of Equation (3) with respect to regularization weighting parameter q.

The unfolded image is stored in an image memory **70**, and can be displayed on a user interface **72**, stored in non-volatile memory, transmitted over a local intranet or the Internet, viewed, stored, manipulated, or so forth. The user interface **72** can also enable a radiologist, technician, or other operator of the magnetic resonance imaging scanner **10**

5   to communicate with the magnetic resonance imaging controller **40** to select, modify, and execute magnetic resonance imaging sequences.

With reference to FIGURE 2, a suitable variable density SENSE k-space sampling profile **80** for each coil is illustrated. The sampling density **80** includes a higher-density portion **82** disposed at or near k-space center which is constant and is not

10   undersampled. This not undersampled higher density k-space portion **82** is reconstructed into a regularization image that includes no folding. Due to the limited k-space range of the k-space portion **82**, the resulting regularization images **62** are low resolution images of the field of view.

The k-space sampling profile **80** also includes undersampled portions **84, 86**

15   disposed away from k-space center and extending to $-k_{max}$ and k_{max}, respectively. In one embodiment, the not undersampled k-space portion **82** spans about one-eighth of the k-space range $[-k_{max}, k_{max}]$ of the variable density SENSE sampling profile **80**. This one-eighth ratio provides good regularization images while still providing a substantial improvement in data acquisition speed. However, other ratios between the central, not

20   undersampled k-space portion **80** and the total k-space range $[-k_{max}, k_{max}]$ of the SENSE sampling profile **80** are also suitable. The portion **82** is at least not undersampled, and is optionally oversampled to provide redundant data. Moreover, if oversampling of the portion **82** is performed, the oversampling optionally includes a non-constant sampling density over some or all of the portion **82**.

25   The k-space sampling profile **80** includes a generally Gaussian sampling density transition between the not undersampled central k-space portion **82** and each of the undersampled outer portions **84, 86**. Other sampling density transitions can also be used, however, such as a linear transition indicated in FIGURE 2 by dashed lines.

In the above-described embodiment, a Cartesian k-space sampling

30   trajectory is used, in which phase-encoded lines are read out, and the central (near k=0) lines are fully sampled and the more peripheral phase encoding lines are spaced apart in accordance with the selected variable density SENSE undersampling. In another

embodiment, two-dimensional SENSE is used, in which the second dimension is typically transverse to the x-y plane. In yet another embodiment, a Cartesian k-space sampling trajectory is used, but the variable density SENSE undersampling is performed in the readout direction rather than in the phase encoding direction. In this embodiment, the parameters k_y and r_y in Equation (2) are suitably replaced by k_x and r_x corresponding to the readout direction. In yet another Cartesian k-space sampling trajectory, SENSE undersampling is performed in both the phase encode and readout directions.

With reference to FIGURE 3, in another embodiment radial sampling trajectories 90 are employed. The radial sampling trajectories 90 converge at or near the center of k-space. This geometry of the radial sampling trajectories 90 provides a higher sampling density region 92 near the center of k-space center due to the convergence of the radial sampling trajectories 90. The higher sampling density region 92 is preferably oversampled or at least not undersampled for each coil, and the higher sampling density region 92 of the imaging data acquired by each coil is suitably reconstructed to form the regularization images 62. While FIGURE 3 shows planar radial sampling trajectories 90, it is also contemplated to employ radial sampling trajectories three-dimensionally, spanning a hemispherical or full spherical solid angle.

With reference to FIGURE 4, in yet another embodiment a spiral sampling trajectory 96 that spirals away from or into k-space center is employed. The illustrated spiral trajectory 96 preferably has a constant pitch **P**; however, an expanding or contracting pitch that changes with distance away from k-space center can also be used. The spiral can be in two or three dimensions. In three dimensions, it can spiral continuously, lie on the surface of concentric spheres, or the like. Also, the spiral can include a plurality of linear segments defining a generally square spiral, hexagonal spiral, dodecagonal spiral, or the like. The constant or increasing spiral pitch geometry provides a higher sampling density region 98 in the vicinity of k-space center. The higher sampling density region 98 is oversampled or at least not undersampled for each coil, and the higher sampling density region 98 of the imaging data acquired by each coil is suitably reconstructed to form the regularization images 62. Other sampling schemes with higher central k-space sampling densities are also contemplated.

The invention has been described with reference to the preferred embodiments. Obviously, modifications and alterations will occur to others upon reading

and understanding the preceding detailed description. It is intended that the invention be construed as including all such modifications and alterations insofar as they come within the scope of the appended claims or the equivalents thereof.